"Express Mail" mailing label number: EV 328 248 039 US

Date of Deposit: January 8, 2004

Our Case No.09800240-0070

IN THE UNITED STATES PATENT AND TRADEMARK OFFICE
APPLICATION FOR UNITED STATES LETTERS PATENT

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TITLE: MULTI-FUNCTIONAL PLASMON-RESONANT CONTRAST AGENTS FOR OPTICAL COHERENCE TOMOGRAPHY

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MULTI-FUNCTIONAL PLASMON-RESONANT CONTRAST AGENTS FOR OPTICAL COHERENCE TOMOGRAPHY

[01] STATEMENT OF ACKNOWLEDGMENT OF GOVERNMENT SUPPORT

[02] This invention was made with Government support under Contract Number NAS2-02057 awarded by the National Aeronautic and Space Administration Ames Research Center and the National Cancer Institute. The Government has certain rights in the invention.

[03] BACKGROUND

[04] When imaging biological tissues, it is often desirable to enhance the signals measured from specific structures. Contrast agents, which produce a strong emission or reflection signal, have been utilized in virtually every imaging modality including ultrasound [1], computed tomography [2], magnetic resonance imaging [3], and optical microscopy [4].

[05] Optical coherence tomography (OCT) is an emerging high-resolution medical and biological imaging technology [5-11]. OCT is analogous to ultrasound B-mode imaging except reflections of low-coherence light are detected rather than sound. OCT detects changes in the backscattered amplitude and phase of light from structures in tissue. This imaging technique is attractive for medical imaging because it permits the imaging of tissue microstructure *in situ*, yielding micron-scale imaging resolution without the need for excision and histological processing. OCT can record structures such as cell membranes, nuclei, and other organelles based on morphology-dependent optical characteristics. Because OCT performs imaging using light, it has a one- to two-order-of-magnitude higher spatial resolution than ultrasound and does not require contact with tissue.

[06] High-quality OCT imaging also depends on the mechanism for optical contrast. Backscatter detection is the most commonly used method of contrast generation; however, this is not necessarily the only way to obtain high contrast.

Spectroscopic OCT techniques have been recently developed and can be used as an additional means for generating contrast in OCT images [12,13]. Whereas standard OCT uses the amplitude (envelope) of the backscattered signal to generate a structural OCT image of tissue, spectroscopic OCT techniques extract data from the interference fringes (carrier) of the backscatter. By taking the Fourier or Morlet-wavelet transform over windowed regions of the interference fringes, variations in the spectrum of the reflected light can be obtained. If endogenous molecules or exogenous contrast agents have strong spectral absorption features within the bandwidth of the OCT source, the three-dimensional spatial distribution of these absorption characteristics can be imaged. Substances which exhibit strong and selective absorption signatures, such as melanin and hemoglobin, can be used to identify cell types. For example, spectroscopic OCT has been used to identify melanocytes in mesenchymal cells from the African frog tadpole [6].

[07] In addition to spectroscopic OCT methods, Doppler OCT can be enhanced by dynamic contrast modulation, based upon changes in the phase (frequency) of the interference fringes. Doppler OCT has been used primarily to detect and quantify blood flow within biological tissue [14,15]. Moving objects within the tissue will also induce Doppler frequency shifts in the background light.

[08] OCT was originally developed and demonstrated in ophthalmology for high-resolution tomographic imaging of the retina and anterior eye [16-18]. Because the eye is transparent and is easily optically accessible, it is well-suited for diagnostic OCT imaging. OCT is promising for the diagnosis of retinal disease because it can provide images of retinal pathology with 10 μm resolution, almost one order-of-magnitude higher than previously possible using ultrasound. Clinical studies have been performed to assess the application of OCT for a number of macular diseases [17,18]. OCT is especially promising for the diagnosis and monitoring of glaucoma and macular edema associated with diabetic retinopathy because it permits the quantitative measurement of changes in the retinal or retinal nerve fiber layer thickness. Because morphological changes often occur before

the onset of physical symptoms, OCT can provide a powerful approach for the early detection of these diseases.

[09] Recently, OCT has been applied for imaging a wide range of nontransparent tissues [6,7,919-20]. In tissues other than the eye, the imaging depth is limited by optical attenuation due to scattering and absorption. A "biological window" exists in tissue where absorption of near-infrared (NIR) wavelengths is at a minimum and light can penetrate deep into highly-scattering tissue [21]. Because optical scattering decreases with increasing wavelength, OCT in nontransparent tissues has routinely used 1.3 μm wavelength light for imaging. In most tissues, imaging depths of 2-3 mm can be achieved using a system detection sensitivity of 110 dB (1 part in 10^{11}). OCT has been applied to image arterial pathology *in vitro* and has been shown to differentiate plaque morphology with superior resolution to ultrasound [7,22].

[010] Imaging studies have also been performed to investigate applications in gastroenterology, urology, and neurosurgery [23-25]. High resolution OCT using short coherence length, short-pulse light sources, has also been demonstrated and axial resolutions of less than 5 μm have been achieved [26,27]. High-speed OCT at image acquisition rates of 4 to 8 frames per second for 500 to 250 square pixel images has been achieved [28]. OCT has been extended to perform Doppler imaging of blood flow and birefringence imaging to investigate laser intervention [14,15]. Different imaging delivery systems including transverse imaging catheters and endoscopes, and forward imaging devices have been developed to enable internal body OCT imaging [29,30]. Most recently, OCT has been combined with catheter-endoscope-based delivery to perform *in vivo* imaging in animal models and human patients [10,31-33].

[011] Apart from medical applications, OCT has been demonstrated as an emerging investigational tool for cell and developmental biology. OCT has imaged the development of numerous animal models including *Rana pipiens* and *Xenopus laevis* (Leopard and African frog), and *Brachydanio rerio* (zebrafish) [34,35]. High-speed OCT imaging has permitted the morphological and

functional imaging of the developing *Xenopus* cardiovascular system, including changes in heart function following pharmacological interventions [36]. High-resolution imaging has permitted the real-time tracking of cell dynamics in living specimens including mesenchymal cell mitosis and neural crest cell migration [37]. OCT is advantageous in microscopy applications because repeated non-invasive imaging of the morphological and functional changes in genetically modified animals can be performed overtime without having to histologically process multiple specimens. The high-resolution, cellular-imaging capabilities suggest that OCT can be used to diagnose and monitor early neoplastic changes in humans.

[012] The ability of OCT to perform optical biopsies, the *in situ* imaging of tissue microstructure at near-histological resolution, has been used to image morphological differences between normal and neoplastic tissue. OCT images of *in vitro* neoplasms of the female reproductive tract [38], the gastrointestinal tract [39], and the brain [25] have been investigated. Optical differences between normal and neoplastic tissue were evident, but primarily for late-stage changes. Still, situations exist where no inherent optical contrast exists between normal and pathologic tissue, such as in early-stage, pre-malignant tumors or in tumors which remain optically similar to normal tissue.

[013] In the past, OCT has found numerous medical and biological applications. However, the imaging technique has relied largely on the inherent optical properties of the tissue to provide contrast and differentiate normal from pathological tissue. Phospholipid-coated perfluorobutane microbubbles (ImaRx Pharmaceutical, Tucson, AZ) have been used as a contrast agent for OCT. Although such microbubbles produce a strong OCT signal, blood and tissue also produce fairly strong OCT signals, and the effects of this contrast agent *in vivo* on the visualization of blood vessels are subtle [40].

[014] Despite the rapidly growing acceptance of OCT in biomedical imaging, there are presently few agents available for enhancing optical contrast. This is partly attributable to the use of NIR wavelengths (>800 nm) that are typically employed in OCT, which are outside the range of most optically active materials.

The keen demand for new optical imaging methods has spurred the development of NIR-active contrast agents [41]. Some of the materials currently under investigation include carbon black, melanin, and colloidal particles [42-44]. NIR fluorescent dyes have been reported to produce detectable emissions at micromolar concentrations, and are potential contrast agents because specific spectral features can be detected using spectroscopic OCT techniques [45,46].

[015] Site-directed hyperthermia has long been considered as an attractive and possibly noninvasive alternative to surgery, in which a localized heat source is directed toward the eradication of diseased tissue. Generation of hyperthermia in biological cells and tissues has been shown to compromise the resistance of tissues to chemotherapy or radiation, or result in necrosis at higher temperatures [47]. Noninvasive hyperthermia can be achieved using nanometer-sized particles; for example, the exposure of a suspension of magnetically active colloids to an AC magnetic field produces a temperature increase due to localized magnetothermal effects, with various power loss mechanisms contributing to heat transfer [48].

[016] SUMMARY

[017] In a first aspect, the present invention is a method of forming an image of a sample that includes forming an image of a mixture by exposing the mixture to electromagnetic radiation in the frequency range of infra-red to ultraviolet light. The mixture contains the sample and plasmon-resonant nanoparticles.

[018] In a second aspect, the present invention is a method of destroying tissue, that includes administering anisotropic metallic nanoparticles to the tissue to form a mixture and subjecting the mixture to electromagnetic radiation.

[019] In a third aspect, the present invention is an improved method of forming an image by optical coherence tomography, that includes exposing a patient to electromagnetic radiation, collecting reflected electromagnetic radiation, and forming an image from the collected electromagnetic radiation. The improvement is drawn to administering anisotropic metallic nanoparticles to the patient to

enhance contrast of the image. Further, the anisotropic metallic nanoparticles include gold nanorods with a magnetic tip.

[020] In a fourth aspect, the present invention is a method of forming an image of a sample that includes forming an image of a mixture by exposing the mixture to electromagnetic radiation. The mixture includes the sample and metallic nanoparticles composed of gold, silver, or copper.

[021] Definitions

[022] The phrase "a compositional modification" means a modification that results in a change in the chemical composition of a nanoparticle, such as the addition of gold to a nanoparticle comprising silver.

[023] The phrase "a surface modification" means a modification that results in the addition of a small molecule or ligand to the surface of a nanoparticle, such as the conjugation of folate to the surface of a gold nanoparticle.

[024] The phrase "enhancing the contrast" means that an image produced with the enhancement shows a greater difference in adsorbed, scattered or reflected electromagnetic radiation between parts of the image, than an otherwise identical image produced without the enhancement.

[025] The term "image" means data produced by receipt of electromagnetic radiation, which may or may not be formed into a picture viewable by the human eye. This includes images produced directly onto a medium such as film or video.

[026] The phrase "infrared to ultraviolet" means electromagnetic radiation having a frequency of 10^{12} to 10^{17} Hz, which excludes radio waves, microwaves, X-rays and gamma rays.

[027] The term "light" means visible light.

[028] The term "nanoparticle" refers to a particle that has a longest axis with a length of at most one micrometer.

[029] The term "nanoellipsoid" refers to a nanoparticle that is shaped like an ellipsoid.

[030] The term "nanorod" refers to a nanoparticle that is shaped like a rod.

[031] The term "nanosphere" refers to a nanoparticle that is spherical.

[032] The term "nanotriangle" refers to a nanoparticle that is shaped like a triangle.

[033] The term "surface plasmons" refers to collective electronic excitations that enhance the optical response of metal particles at a frequency (*e.g.*, displaying an extinction coefficient of at least 10^6 $M^{-1}cm^{-1}$).

[034] The phrase "plasmon-resonant nanoparticles" refers to metallic nanopaticles that have an extinction coefficient of at least 10^6 $M^{-1}cm^{-1}$ at some frequency in the range of 10^{12} to 10^{17} Hz.

[035] The phrase "plasmon-resonant contrast agent" refers to plasmon-resonant nanoparticles suitable for use as a contrast agent to form images of biological tissue using optical coherence tomography, light microscopy, holography, confocal microscopy, polarization microscopy, interference microscopy, multi-photon microscopy, and endoscopy.

[036] BRIEF DESCRIPTION OF THE DRAWINGS

[037] Figure 1 depicts wavelength specific plasmon-resonances calculated for gold nanorods with the indicated aspect ratios;

[038] Figure 2 depicts bimetallic Ag-Au nanorod synthesized by sequential electrodeposition into porous template;

[039] Figure 3 depicts two-step anodization and pore widening process for nanoporous alumina templates;

[040] Figure 4 depicts resorcinarenes as nanoparticle dispersants;

[041] Figures 5A and 5B depict Au nanorods prepared by electrodeposition (width: 75 nm; aspect ratio: 10:1 and 4:1, respectively); Figure 5C depicts Au nanorods prepared by chemical reduction (width: 25 nm; aspect ratio: 10:1);

[042] Figure 6 depicts ligand-functionalized nanoparticles via tandem surfactant cross-linking/ligand cross-metathesis (folate is denoted as "Fol");

[043] Figure 7 depicts a synthetic scheme of resorcinarenes bearing substituents **5** and **6**; and

[044] Figure 8 depicts a synthetic scheme of cross-metathesis of ligand onto cross-linked surfactant model.

[045] DETAILED DESCRIPTION

[046] The present invention makes use of the discovery that plasmon-resonant nanoparticles can be used to enhance the contrast in analyses and imaging techniques that use electromagnetic radiation, particularly those techniques which use radiation in the frequency range of infrared to ultraviolet, such as optical coherence tomography, light microscopy, holography, confocal microscopy, polarization microscopy, interference microscopy, multi-photon microscopy, and endoscopy. Moreover, metallic nanoparticles composed of gold, silver, and/or copper are particularly suited as contrast agents for OCT applications. Preferably, the nanoparticles are metallic anisotropic nanoparticles, which possess superior plasmon-resonant characteristics and may be fabricated in bimetallic forms to permit their use in OCT applications using switchable magnetic and electric fields. Furthermore, the nanoparticles can be functionalized with biomolecular ligands for cell-specific delivery to permit site-directed OCT imaging. The nanoparticles efficiently absorb the incident optical radiation and can be used as hyperthermia agents, creating local thermal gradients that are sufficient to kill individual cells. These contrast agents can therefore be used simultaneously for the detection and imaging of targeted cells followed by hyperthermic ablation.

[047] A nanosphere represents one form of plasmon-resonant nanoparticle that is useful in the present invention. Preferred nanospheres have diameters of 10 nm to 1 μm. More preferably, the nanospheres have diameters of 20 nm to 500 nm. Even more preferably, the nanospheres have diameters of 40 nm to 200 nm. Methods for the synthesis and modification of nanospheres have been described in the art [88-93,112].

[048] Colloidal gold nanoparticles are commonly used in biological and biomedical applications because of their inertness under physiological conditions. They are also well known for their intense absorption and scattering properties. Gold nanoparticles functionalized with biomolecular ligands have been employed

as carriers and labels in biological tissue staining [51], drug and gene delivery [52-54], and biosensing applications [55]. In this regard, any metallic particle may be coated with gold. The optical responses of colloidal gold particles are enhanced by collective electronic excitations known as surface plasmons, which are responsible for extinction coefficients in the range of 10^9–10^{11} M^{-1} cm^{-1} [56]. Gold nanoparticles can have anisotropic absorption properties which vary with respect to their orientation relative to the incident optical radiation. Plasmon-resonant nanoparticles can thus be detected at extremely low concentrations, with several orders of magnitude greater sensitivity than organic dye molecules [57]. Furthermore, their optical emissions do not bleach over time and have no saturation limits; in other words, they are optically indefatigable.

[049] Recent activities in nanoscale materials science have further expanded the knowledge base on the optical physics of metallic nanoparticles, and it is now evident that physical structure has a dramatic influence on plasmon-enhanced response. In particular, the optical resonances of anisotropic metallic nanoparticles such as rods, ellipsoids, and triangles have been found to be more intense and frequency-specific than their spherical counterparts, and can be tuned as a function of their size, shape, and interparticle coupling [49,50]. These variables provide a wealth of opportunities for designing contrast agents with enhanced optical properties at specific wavelengths. Furthermore, metallic anisotropic nanoparticles that possess electromagnetic properties (*e.g.*, bimetallic nanoparticles) permit their use in OCT applications where their orientation can be modulated by either magnetic or electric fields.

[050] Anisotropic nanoparticles of a variety of distinctive shapes are useful as contrast agents in optical coherence tomography. Although preferred embodiments demonstrate the synthesis and utility of metallic nanorods, metallic anisotropic nanoparticles composed of other shapes, like triangles and ellipsoids, may be used. These alternative shapes will yield plasmon-resonant characteristics that are suitable for use as OCT contrast agents and hyperthermic ablation applications.

[051] The general fabrication scheme for preparing other metallic anisotropic nanoparticles differing in shape (*e.g.*, nanotriangles and nanoellipsoids) is an extension of the types of fabrication techniques that are disclosed in the embodiments for metallic nanorods. Nanoporous alumina templates can be prepared that possess unique shapes using controlled etching processes. These templates can then be backplated with metallic Ag, followed by electrodeposition of Au at constant current using a standard gold electroplating solution. Nanoparticles are subsequently released into solution by dissolution of the silica template and recovered by centrifugation. Controlled deposition times can afford anisotropic nanoparticles of various sizes.

[052] Preferably, the nanoparticles are metallic. Preferred metals include gold, silver, copper, cobalt, nickle, iron, and alloys or mixtures thereof. Metallic nanoparticles permit further compositional modification or surface modification of the particles. For example, a compositional modification of a non-magnetic nanosphere (*e.g.*, a gold nanosphere) can be accomplished using cobalt, iron, or nickel to produce particles with magnetic properties. Such magnetic metallic nanoparticles are useful plasmon-resonant contrast agents in applications where OCT is conducted with switchable electric or magnetic fields. Likewise, a surface modification of a metallic nanoparticles (*e.g.*, gold nanoparticles) can be accomplished with small molecules (*e.g.*, crosslinking agents), ligands (*e.g.*, folate) or receptors (*e.g.*, antibodies). Functionalized nanoparticles are useful in the present invention for targeted delivery of nanoparticles to specific cell types (*e.g.*, cancer cells). Furthermore, the nanoparticles may be present as part of a composite. The composite may contain other materials which hold a number of the nanoparticles together, for example ceramics such as alumina, silica or glass; and organic materials such as proteins, lipids, polymers, and carbohydrates.

[053] The development of metallic nanorods as contrast agents imparts several unique parameters for enhancing adsorption or backscattering detection by OCT. First, the nanorods can be optimized for extinction at NIR frequencies as a function of aspect ratio, independently from their diameters; second, the size of their absorption or scattering cross sections can be engineered as a function of unit

particle volume; third, the anisotropy of plasmon-enhanced optical response is sensitive to the relative angle of incidence; and fourth, the ability to synthesize bimetallic nanorods allows one to couple plasmon-enhanced optical properties with magnetic properties for novel contrast enhancement.

[054] Surface plasmons in metallic nanoparticles are generated by the collective excitation of free conduction electrons in response to a characteristic optical frequency. An important feature of plasmon resonance is its high sensitivity to shape anisotropy: isolated spherical nanoparticles typically support a single resonance frequency, whereas anisotropic particles (rods, triangles, ellipsoids, *etc.*) will exhibit at least one additional plasmon mode [56]. In the case of cylindrical nanorods, the frequency of this second (longitudinal) plasmon mode is determined primarily by the particle's aspect ratio, and is redshifted well into the NIR. It has been shown, both theoretically and experimentally, that gold nanorods with aspect ratios of 4:1 exhibit longitudinal plasmon resonances centered at 800 nm, whereas nanorods with aspect ratios of 9:1 exhibit resonances centered at 1.3 μm [49,50,58] (*see* Figure 1).

[055] Methodologies for synthesizing metallic nanorods are now well established. Nanoparticle-seeded growth mediated by cationic surfactants can produce cylindrically symmetric nanorods with aspect ratios as high as 20:1, with diameters on the order of 10–20 nm [59,60]. For thicker nanorods (>20 nm), pulsed electrodeposition into metallized nanoporous membranes has been demonstrated to produce nanorods of nearly every aspect ratio [58,61,62]. The latter synthetic method offers excellent control over nanorod dimensions: rod thickness is predetermined by pore diameter, whereas rod length is a direct function of deposition time. In addition, different metals can be deposited sequentially, permitting the synthesis of bimetallic nanorods with dual materials properties (*see* Figure 2).

[056] Nanorods can be prepared with preferred aspect ratios of 1.5:1 to 20:1. More preferably, nanorods can be prepared with aspect ratios of 4:1 to 9:1, which display resonant extinction peaks at 800 nm and 1.3 μm, respectively. Nanorods can be prepared with diameters of 10 nm to 300 nm. More preferably, nanorods

can be prepared with diameters of 10 nm to 100 nm. The synthesis of metallic nanorods has been optimized by the electrochemical reduction of metallic salts inside nanoporous alumina membranes, as well as by chemical reduction in micellar solutions. The latter method is useful for synthesizing nanorods with diameters of 10 to 20 nm, which can be tuned with a narrow size dispersity by varying the particle seed size, the surfactants comprising the micellar reaction template, or the concentration of associative electrolytes serving as "capping agents" [59,60]. Electrodeposition can be used to make nanorods of 20 to 300 nm in diameter [58,61,62]; in this case, the nanoporous template controls the size and quality of the nanorod dispersions. Individual nanorods of 20 to 40 nm in width have strong absorption and moderate scattering efficiencies, while nanorods of 50–100 nm in diameter have intense NIR scattering cross sections. Transverse plasmon modes also produce strong optical responses in the visible range, regardless of aspect ratio; for example, both nanorods and nanowires (d~30 nm) have been shown to be efficient scattering agents near their transverse plasmon resonance (λ_{max}~530–580 nm) [63].

[057] It is preferrable to obtain nanorods with narrow size dispersity (<10%). With respect to micelle-templated synthesis, the nanorods must be separated and refined by size-selective precipitation or other means of separation. Nanorods can be isolated by size-exclusion techniques assisted by centrifugation, since this technique has been successful in enriching the population of anisotropic nanoparticle oligomers [64]. With respect to electrochemical synthesis, it is essential to produce nanoporous templates of high quality in order to obtain nanorods of narrow size dispersity. Commercially available alumina membranes often have nonuniform or poorly defined pore sizes; therefore, one can prepare nanoporous membranes with customized diameters by anodizing high-grade aluminum (99.9999%, PVD Materials) under well-defined conditions. Highly uniform nanoporous alumina can be prepared by a two-step anodization process: (1) a sacrificial oxide layer is first formed and removed by hydrolysis, leaving behind a periodically dimpled surface; (2) a second anodization produces a periodic array of nanopores, whose diameters are determined by the applied

voltage [65-67]. The pores can also be further enlarged by acid-assisted electrochemical etching [68]. This process has been used to prepare alumina membranes with 60- and 90-nm pores (*see* Figure 3).

[058] These and other methods for preparing nanorods are available to one of ordinary skill in the art and have been described in the literature [59-63,69,70]. Likewise, general procedures for the preparation of nanotriangles are disclosed in the literature [71-73].

[059] Biomolecular ligands can be conjugated onto colloidal particles by simple electrostatic adsorption [51]. Protein-conjugated nanoparticles can also be internalized by receptor-mediated endocytosis [74-78], thereby providing a useful mechanism for cell uptake. Electron microscopy studies demonstrate that nanoparticle uptake correlates strongly with ligand-induced receptor clustering [79,80], and that the conjugated proteins are densely packed on the particle surface [81]. These have obvious applications for site-directed OCT imaging as well as for localized hyperthermia.

[060] Nanoparticles have also been functionalized with small-molecule ligands, most notably by the spontaneous chemisorption of thiols onto gold or gold-coated nanoparticles [82]. However, the adsorption of thiolated ligands is reversible and results in facile surface exchange. In other embodiments, surface functionalization of gold nanoparticles can be achieved through chemical modification [82-94] and through use of biomolecule ligands [74-81,95]. The methods for coating metallic nanoparticles in chemically robust, long-lived shells, some of which are amenable to chemical functionalization, are generally available to one skilled in the art [88-93].

[061] Passivating organic surfactants such as *n*-alkanethiols have been widely used to functionalize and disperse metallic nanoparticles. However, such surfactants are prone to surface exchange, which limits their robustness as ligands for biological applications. The dispersion and functionalization of colloidal metallic nanoparticles in the 10–200 nm range are accomplished by developing surfactants from a class of macrocycles known as resorcinarenes [84-86,96-98]. These compounds possess large, multivalent headgroups for robust adsorption

onto nanoparticle surfaces, and several hydrocarbon tails per molecule spaced several angstroms apart (*see* Figure 4). The hydrocarbon tails ensure a high degree of configurational freedom per chain in the surfactant layer, which translates into effective dispersion control.

[062] Dispersion studies have been performed using Au nanoclusters encapsulated by resorcinarenes with polyoxygenated headgroups (*see* Figure 4, compounds **1** and **2**) [84,98] and by resorcinarenes with tetrathiol headgroups (*see* Figure 4, compounds **3** and **4**). The latter are able to extract colloidal Au particles as large as 100 nm from aqueous suspensions and disperse them into organic solvents for further chemical functionalization [85,86]. Preferred nanoparticles include those that are encapsulated by compounds **3** or 4, as they are highly resistant to surface exchange, and are capable of resisting degradation by competing surfactants and other adsorbates over a period of several weeks.

[063] These and other techniques for dispersing nanoparticles using resorcinarenes are described in U.S. Patent application Serial No. 10/218,185 entitled "NANOPARTICLE ARRAYS AND SENSORS USING SAME," to Alexander Wei *et al.*, filed on August 12, 2002 and published April 10, 2003, the contents of which are hereby incorporated by reference.

[064] Optical hyperthermia can be induced at both visible and NIR wavelengths. Fully noninvasive optical hyperthermia is possible if two criteria can be met: (1) a mechanism for site-specific energy delivery to the target area with minimal collateral damage, and (2) energy transport such that the intervening tissue is not subject to thermal heating. The latter can be achieved using ultrashort laser pulses, which deliver coherent light to tightly confined focal regions in high-energy bursts. By concentrating the laser energy into a train of high-peak-power pulses, the average power can be decreased and collateral heating of surrounding tissue can be minimized. This effect has been studied in retinal tissue, in which linear increases in pulse power produced nonlinear increases in tissue damage [99-101]. By developing nanoparticles with high NIR responses as site-directed multifunctional agents, the primary function of OCT as an imaging technology is thereby extended to yield a complementary therapeutic technique.

[065] OCT has been applied toward image-guided laser ablation of surgical tissue *in vitro* [102]. Laser-induced hyperthermia was performed using a high-power argon laser (514 nm) to thermally coagulate blood within two vessels, located 1.5 mm deep in muscle tissue. Coagulation within the blood vessels (0.0–0.5 sec) occurred prior to overlying tissue damage, due the high absorption coefficient of blood at 514 nm. The image sequence demonstrates OCT monitoring of the process in real time, providing feedback to the surgeon. In this case, optical imaging and laser-induced hyperthermia were performed using two separate instruments. Optionally, OCT with variable incident optical power can be used in conjunction with NIR-resonant nanorods for *in vivo* targeting and noninvasive destruction of cells deep within tissue.

[066] EXAMPLES

[067] *Example 1. Synthesis of gold nanorods*

[068] Commercially available nanoporous alumina templates (Anodisc membranes, Whatman) were backplated with metallic Ag, followed by electrodeposition of Au at constant current (0.6 mA/cm^2) using a standard gold electroplating solution (Technics, Inc.). Nanorods were released into solution by dissolution of the alumina template and recovered by centrifugation. Controlled deposition times afforded nanorods of various aspect ratios ranging from 4:1 to 10:1 (*see* Figure 5A,B). The optical properties of these nanorods extend well into the NIR, and can be tuned for extinction at precise wavelengths as a function of aspect ratio. We have also prepared gold nanorods with much narrower widths by surfactant-mediated synthesis in aqueous solutions. Small (3-4 nm) gold particles were used to nucleate the reduction of gold chloride by ascorbic acid in the presence of cetyltrimethylammonium bromide, following the protocol of Murphy and coworkers [60]. The aspect ratio of the gold nanorods was controlled by adjusting the seed solution amount; in the example above, nanorods were produced with an aspect ratio of about 10:1 (*see* Figure 5C).

[069] *Example 2. Optical characterization of tissue models*

[070] Tissue models consisting of a mixture of dyes for absorption, polystyrene latex microspheres for scattering (Bangs Laboratories), and agarose gelatin for structural support (Sigma Chemicals) will be constructed to simulate the optical characteristics of different tissues [103]. Absorption and scattering coefficients can be independently controlled to represent various tissue types. Absorption properties may be modified by the addition of mixtures of FD&C Yellow #5 and FD&C Blue #1. These dyes increase absorption preferentially in the 630–700 nm region of the optical spectrum. Although the addition of these dyes will simulate tissue optical properties, they have minimal interference with OCT imaging which will be performed at NIR wavelengths (800-1300 nm). Therefore, imaging characteristics will be strongly dependent on the scattering properties of the tissue model versus the absorption properties. Scattering properties can be easily adjusted by varying the concentration and size of polystyrene microspheres [103].

[071] Nanorods may be incorporated into the tissue model in a variety of ways. The optical contrast agents may be added in as layers at varying concentrations and depths. Optionally, the nanorods may be suspended in a liquid solution and drawn up into a 1-cc syringe with a 28-gauge needle in order to introduce contrast agents into a spatially localized region. This may be positioned over the tissue model using a 3-axis micromanipulator stage, enabling contrast agents to be injected at a precise location. The absorption and scattering properties of tissue models may be determined with and without the addition of optical contrast agents using oblique-incidence optical fiber reflectometry [104,105]. This technique is a simple and reliable method that can be readily incorporated into the OCT instrument and used for tissue models, *in vitro* cell cultures, and *in vivo* tissues.

[072] The absorption and reduced scattering coefficients of the sample can be determined by viewing the OCT imaging beam on the surface of the sample or specimen with a CCD camera sensitive to NIR wavelengths (Princeton Instruments). Prior to OCT imaging, the tissue model or *in vivo* specimen may be placed on the OCT instrument stage so that the incident beam is at a known oblique angle. The CCD camera will capture the diffuse reflectance pattern of the

OCT beam on the sample or specimen. From this image, one can determine the distance between the point of incidence and the apparent center of diffuse reflectance. Using this measurement and the known angle of incidence, the absorption and reduced scattering coefficients can be calculated [104,105]. For tissue models or *in vivo* specimens that have localized regions of contrast agents, two-dimensional (2D) maps may be generated to represent the spatial distribution of the optical properties (absorption and reduced scattering coefficients). Maps may be generated by stepping the position of the sample or specimen using computer-controlled stages, repeatedly capturing CCD images and calculating the optical properties.

[073] *Example 3. Synthesis of magnetically tipped nanorods*

[074] Plasmon-resonant nanorods appended with magnetic tips may be synthesized by sequential electrodeposition of Au and Ni in nanoporous alumina templates, as previously described [62]. Ni is preferred for several reasons: first, it can be electrodeposited at low voltages; second, its high magnetic shape anisotropy and low crystalline anisotropy make it easy to predict and control nanorod orientation in an applied magnetic or electric field; and third, it forms a stable, hydrophilic oxide shell which is resistant to degradation under physiological conditions. Although Ni is less desirable from a toxicological perspective, the amounts involved are submicrogram in quantity, and other magnetic materials can be used. One may prepare Au/Ni nanorods with a [4+2]:1 aspect ratio; the Au segment will be resonant at 800 nm, and the Ni segment exhibit a uniaxial magnetic dipole. Although Ni and Au can be deposited in either order, unreduced metal salts from the first deposition residing in the pores can affect the final composition; therefore, TEM analysis may be used for accurate structural characterization.

[075] *Example 4. Oriented and dynamic optical response of magnetically and electrically active nanorods (prophetic example)*

[076] The modulation of the optical responses of the bimetallic Au/Ni nanorods will be evaluated using applied magnetic fields for orientation effects, at both constant and time-modulated (dynamic) field strengths. Using the solenoid integrated with a microscope, a static magnetic field will be applied to the tissue model with the magnetic field vector aligned either at 0° or 180° with respect to the direction of the incident optical beam. This will cause the polar magnetic nanorods to align with the magnetic field and change the optical scattering and absorption properties compared to the randomly oriented state. Quantitative measurements will be made, along with measurements from field orientations in the range of 0° to 180° to determine the relative backscatter and absorption properties of the nanorods with respect to orientation angle. These physical measurements will help one skilled in the art to develop a better understanding of the novel magneto-optical contrast mechanisms observed so far and a rational basis for further optimization.

[077] In a similar manner, the electric dipoles of the magnetic bimetallic particles will be oriented as a function of electric field strength in switchable electric fields.

[078] *Example 5. In vitro optical modulation by magnetic particles (prophetic example)*

[079] Magnetic particles confined within intracellular compartments will be detected optically under externally applied fields. Macrophages (ATCC #TIB-67) will be cultured overnight in the presence of three forms of magnetic colloids or suspensions, which were internalized by phagocytosis. The samples that will be studied include: (1) ferromagnetic hematite (Fe_2O_3) nanoparticles aggregated into micron-sized clusters suspended in phosphate-buffered saline (PBS), (2) superparamagnetic magnetite (Fe_3O_4) colloid, and (3) super-paramagnetic latex beads (40% magnetite + 60% divinylbenzene; 2.38 μm in diameter) suspended in PBS. Cell counts after incubation for 48 hours should indicate healthy cultures for all samples. Dishes containing the magnetically labeled macrophages will then

placed atop a solenoid coil on a bright-field microscope, and monitored with a water-immersion objective under an applied magnetic field (~300 G). The corresponding gradient in the axial direction should be estimated to be ~6 T/m at the dish surface.

[080] Cells containing the magnetic nanoparticles will be monitored under both constant and fluctuating magnetic fields. Intracellular movement should be detectable with particles (1) but not with particles (2) or (3), suggesting that ferromagnetic behavior was important for this form of contrast. Translational movement of the hematite nanoparticles will be monitored in both the axial and transverse directions; rotational alignment of the particles will also be monitored as the magnetic field was switched on.

[081] *Example 6. Functionalization of nanorods with bioreceptors (antibodies) (prophetic example)*

[082] The cell-surface folate receptor will be chosen as a target for nanorod labeling and OCT contrast detection. Nanorod agents will be functionalized using monoclonal antibodies raised against folate receptors (human folate receptor a: MOV18/ZEL, Alexis Biochemicals; folate-binding protein: ab2107, abcam) as well as the folate ligand itself (see below). Methods for conjugating antibodies onto colloidal gold nanoparticles ("ImmunoGold") are well-established and will be applied toward nanorod contrast agents [51]. The antibody-conjugated nanorods will be typically suspended at a concentration of 10^{10} particles/mL, and administered *in vivo* for site-directed labeling of carcinoma cells in the hamster cheek pouch model (see below).

[083] *Example 7. Functionalization and coating methodologies for ligand-conjugated nanorods (folates) (prophetic example)*

[084] Nanorods will be encapsulated in crosslinked surfactant layers. This surface is ideal for grafting complex organic ligands such as folate using olefin cross-metathesis [106,107]. Encapsulated nanorods will be isolated from excess surfactant by centrifugation, then resuspended and treated with

$Cl_2(Cy_3P)_2Ru$=CHPh at sub-millimolar concentrations. These will be centrifuged again and redispersed in the presence of allyl folate ester. Ru-catalyzed olefin metathesis is a well-known "living" polymerization method, and is tolerant of an enormously diverse range of functional groups and reaction conditions [108]. Some of the catalytic Ru will be covalently tethered to the nanorod surfactant layer, with additional coordination by nearby crosslinked *cis*-olefins. These *cis*-olefins are capable of further metathesis with the Ru catalyst, so that the degree of intermolecular crosslinking within the resorcinarene monolayer ("annealing") will gradually increase over time. A modification of this approach will permit the addition of allyl folate to the Ru-conjugated nanorods without removing excess Ru catalyst to yield Ru=alloxylidene folate at submillimolar concentrations. These will undergo cross-metathesis with *cis*-olefins in the crosslinked shell to yield folate-labeled nanorods with high surface coverage.

[085] *Example 8. Nanoparticles stabilized by cross-linked surfactant shells*

[086] Resorcinarene-encapsulated nanoparticles are robust ligand-directed agents for imaging and therapeutic applications. While surfactants bearing substituents **3** and **4** have superior resistance against desorption, long-term stability and chemical erosion remain critical issues for biological applications. To overcome these issues, a surface polymerization protocol was developed that crosslinks resorcinarene surfactant layers into robust yet functional coatings (*see* Figure 6). Nanoparticles are encapsulated by surfactants with terminal alkenes such as substituents **5** and **6**, which can be cross-linked by olefin metathesis using $Cl_2(Cy_3P)_2Ru$=CHPh, a ruthenium carbene catalyst introduced by Grubbs [106,107]. The resorcinarene coatings are sufficiently dense to allow for significant intermolecular cross-linking, thereby yielding a nondesorptive surfactant monolayer. The cross-linked shells are capable of further metathesis with alkene-bearing molecular ligands, so that their covalent attachment is irreversible.

[087] A test study was conducted using Au nanoclusters encapsulated by resorcinarene tetraolefin substituent **5** [109]. Kinetic studies on the intramolecular

metathesis of **5** by $Cl_2(Cy_3P)_2Ru{=}CHPh$ using ^1H NMR spectroscopy provided effective first-order rate constants at different catalyst loadings and reactant concentrations, which suggested reaction conditions for producing surfactant layers with a high degree of crosslinking. Dispersions of encapsulated Au nanoclusters treated in this manner were indeed found to be highly robust; however, nanoparticles encapsulated in highly crosslinked shells were poorly dispersible because of the loss of surfactant chain mobility, which limited their processing potential. A second-generation system based on tetrathiol–tetraolefin substituent **6** has been developed, in which strong chemisorption and olefin metathesis operate synergistically to produce highly robust but dispersible encapsulated nanoparticles. Resorcinarene bearing substituent **6** was synthesized according to Figure 7 ((a) HCl, EtOH, 65 °C (70% yield); (b) i) *N*-bromosuccinamide, 2-butanone; ii) CH_2BrCl, Cs_2CO_3, DMF, 100 °C (40% yield over 2 steps); (c) *n*-BuLi, THF, -78 °C to 0 °C (85% yield)) and used to extract 20-nm colloidal Au particles from aqueous solutions into toluene. Cross-linking was performed by treating nanoparticle suspensions with Ru metathesis catalyst for 5 minutes (5×10^{11} particles/mL, or ~1 pM), then quenching with ethyl vinyl ether. Tetrathiol substituent **6** did not have a negative impact on the catalyst's activity, such that cross-linking could be performed in the presence of excess surfactant.

[088] For these and other techniques for dispersing nanoparticles using resorcinarenes, see U.S. Patent application Serial No. 10/218,185 entitled "NANOPARTICLE ARRAYS AND SENSORS USING SAME," to Alexander Wei *et al.*, filed on August 12, 2002 and published April 10, 2003.

[089] The robustness of the cross-linked surfactant shells can be evaluated by exposing the nanoparticle dispersions to dodecanethiol, a strongly adsorbing surfactant with poor dispersant properties. All resorcinarene-stabilized dispersions were observed to be stable after a one-week exposure to dodecanethiol at room temperature; however, those which were not subjected to metathesis-mediated crosslinking degraded within a few hours at 70 °C. Nanoparticle dispersions treated with 0.01 mM catalyst or less were also degraded within a few hours, but nanoparticles treated with 0.1 mM catalyst were found to be completely stable

under these conditions. Nanoparticles stabilized in cross-linked surfactant layers could also be repeatedly precipitated by centrifugation and redispersed by mild sonication with high levels of recovery.

[090] These experiments demonstrate that shells of cross-linked **6** can be highly resistant to surface desorption, while retaining excellent dispersibility in various solvents for subsequent functionalization. Transmission electron microscopy (TEM) provides further evidence correlating dispersion stability with metathesis cross-linking; uncross-linked nanoparticles precipitated by dodecanethiol were redispersed and cast onto Formvar-coated TEM grids, and were observed to have formed aggregates. Nanoparticle dispersions treated with low catalyst loadings produced mixtures of monodispersed and aggregated particles, whereas dispersions treated with high catalyst loadings were completely redispersed.

[091] *Example 9. Functionalization of crosslinked nanoparticle shells*

[092] Nanoparticles stabilized in monolayer shells of **6** were treated with Ru metathesis catalyst, then centrifuged without quenching and redispersed in fresh solvent containing polymerizable norbornene derivatives or alkene-bearing ligands. An inverse approach to cross-metathesis–crosslinking may be used wherein which the Ru catalyst is first treated with an alkenyl ligand such as substituent **7a** or **7b** (see Figure 8 ("NPhth" denotes phthalimide, (a) **7a** or **7b** (1 equivalent), $Cl_2(Cy_3P)_2Ru{=}CHPh$ (1 equivalent), tetrabromoresorcinarene cis-deine (2 equivalents), toluene)). Cross-metathesis of the resulting Ru=alkylidene intermediate with resorcinarene *cis*-diene substituent **8** (a model compound for the crosslinked surfactant layer) produces a 1:1 adduct in significant quantities along with some doubly functionalized resorcinarene; the cross-coupling reaction between substituents **7b** and **8** proceeds in similar fashion. These reactions are the molecular analogy to the proposed surface functionalization, and support the notion that alkenyl ligands can be reliably grafted onto the crosslinked surfactant layer.

[093] *Example 10. Evaluation of anisotropic nanoparticle uptake by cells in vitro and intracellular nanorod aggregation (prophetic example)*

[094] Site-directed nanorod labeling and possible cell uptake will be evaluated with *in vitro* cultures of KB tumor cells, which express high levels of folate receptor on their surfaces [110]. For example, cells will be plated and exposed to antibody- and folate-conjugated nanorods at femtomolar to low picomolar concentrations to establish their relative affinity of binding. Cell labeling will be characterized first by OCT imaging, then by cryo-electron microscopy and by a standard silver-staining histological protocol [51]. Antibody-labeled nanorods should bind to the exterior of KB cells, whereas folate-labeled nanorods should be taken up by receptor-mediated endocytosis (cf. Figure 3a). This will provide an important variable in optical contrast generation, as the latter nanorods exist in an aggregated state. Differences in optical contrast quality will determine whether antibody-labeled nanorods are preferred over folate-labeled nanorods, or vice versa.

[095] *Example 11. In vivo OCT imaging of hamster cheek pouch tumors using folate-labeled nanorods (prophetic example)*

[096] The Syrian Golden hamster cheek pouch carcinogenesis model closely resembles the events involved in the development of premalignant and malignant human oral cancers [108,110,111]. This is a well-characterized model for squamous cell carcinoma, which is the leading form of malignancy in human skin, oral, and genital mucosas. The cheek pouch model is ideal for imaging and contrast localization studies on pre-malignant through metastatic stages.

[097] Syrian Golden hamsters (100 to 120 g) will have cotton sutures inserted submucosally into both cheek pouches under intraperitoneal anesthetics (pentobarbital sodium, 500 mg/kg). A 0.5% solution of 9,10-dimethyl-1,2-benzanthracene (DMBA, Sigma Chemical) will be painted on the left cheek pouches biweekly, starting 4 weeks after the sutures are placed; the right cheek pouch will serve as the normal control. Imaging of tumor sites in the hamster cheek pouch will take place weekly, beginning at 10 weeks after tumor induction

and while the tumors are at early pre-malignant stages. At early stages, abnormal dysplastic cells are likely to be present before changes can be observed visually. Four animals will be imaged each week; prior to imaging, animals will be anesthetized with intraperitoneal anesthetics (pentobarbital sodium, 500 mg/kg). The cheek pouch will be everted and the animal will be positioned on the instrument stage under the OCT imaging beam. Following CCD image collection to measure the optical properties of the *in vivo* tissue, 3D OCT imaging will be performed at the site indicated by the suture. Both the left (tumor site) and the right (control site) cheek pouches will be imaged. Subsequently, a solution of freshly suspended functionalized nanorods ($\sim 10^{10}$ nanorods/μL) will be injected intravenously via a tongue vein. Doses will range of 5 μL to 100 μL, depending on the efficiency of uptake by the neoplastic cells.

[098] Thirty minutes following injection, 3D optical property measurements and OCT imaging will be repeated for both cheek pouches. If magneto-optical contrast mechanisms are to be employed, then animals will be positioned on the OCT imaging stage with the solenoid positioned over the tumor sites to orient the magnetic nanorods at varying angles relative to incident optical radiation. OCT imaging will be performed in different experiments at 800 nm and 1.3 μm, and for varying magnetic field orientations (0° to 180° with 10° increments). For each series of experiments, real-time structural and spectroscopic OCT imaging will be used to detect changes in the scattering and absorption properties of the tissue. Correlations will be made between the 2D maps of optical properties and the 3D OCT images. Additionally, measurements from OCT images will characterize the enhancement in tumor detection by the contrast agent.

[099] At the conclusion of each weekly imaging experiment, one of the animals will be allowed to recover. The three remaining animals will be euthanized by CO_2 inhalation, and tissue from the cheek pouches will be excised for histopathological observations using light, confocal, and electron microscopy. Microscopy findings of tumor morphology and contrast agent localization will be

correlated with the OCT image findings to determine specific OCT image features that indicate the presence of the nanorod contrast agents.

[0100] *Example 12. Optical hyperthermia on tissue phantoms with real-time OCT monitoring (prophetic example)*

[0101] Optical hyperthermia studies will be performed in tissue models with and without plasmon-resonant contrast agents. Structural and spectroscopic OCT imaging of tissue models will be performed using low (5 mW) optical power using optimized sampling configurations and contrast mechanisms. OCT will then be performed using incrementally larger optical powers (5 mW steps, up to 50 mW) while observing for signs of hyperthermia-induced changes in real time. In case the lowest incident power is still too high, off-resonant nanorods with different aspect ratios will be used to attenuate their plasmonic responses. Imaging exposure time of the tissue model will initially be 1 minute at each optical power level, and will be varied from 1 to 5 minutes to determine exposure thresholds for hyperthermia-induced changes. Using OCT, changes are likely to include local increases in optical backscatter and structural changes in the tissue model. Due to the presence of the highly absorbing nanorods, local hyperthermia effects should coincide with the spatial distribution of the nanorods as detected using OCT. Simultaneously with real-time OCT imaging, one may record local temperature changes within the tissue model using a miniature thermistor embedded within the agarose layers of a tissue model and record surface and bulk temperature changes using a temperature-calibrated infrared camera. Following each imaging session, the tissue model will be physically sectioned along the OCT imaging plane using a razor blade and placed on a glass microscope slide for high-resolution light microscopy of the imaging site. Digital light microscopy will be used to verify the presence or absence of any local hyperthermia effects. These images may be compared with structural and spectroscopic OCT images to determine the threshold at which OCT can detect the onset of hyperthermia.

[0102] *Example 13. Nanoparticle-assisted optical hyperthermia in vivo (prophetic example)*

[0103] Optically-induced hyperthermia data obtained from the tissue model studies will serve as a guide for inducing hyperthermia *in vivo* in animal tumor models. Nanoparticles with folate ligands will be injected intravenously into the tongue vein of an anesthetized hamster with squamous cell carcinomas present in the cheek pouch as previously described. Thirty minutes following injection, animals will be positioned on the OCT imaging stage under optimized imaging configurations. OCT imaging will be performed at 800 nm and 1.3 μm for varying exposure durations (1–5 min at 1-min increments) and optical powers (5–50 mW at 5-mW increments). For each series of experiments, real-time OCT imaging will be used to monitor hyperthermia-induced changes in depth, and infrared camera imaging will be used to detect hyperthermia-induced changes at the surface. Changes indicative of hyperthermia-induced injury include increased optical backscatter or absorption in microstructural or spectroscopic OCT, and changes in the birefringent properties of the tissue (most notably light–dark banding in OCT images from ordered muscle fibers).

[0104] *Example 14. Fabrication of microparticles with a gold coat*

[0105] BSA is also known to interact very strongly with gold nanoparticles via its amine and thiol residues [113]. Because of the presence of thiol and amine groups, the microparticle can be used as a template for the adhesion of gold nanoparticles. Addition of positive polymer layer was not performed for adhesion of the gold colloid unto the surface of the microparticle. The gold colloid can adsorb directly onto the protein shell via interaction with thiol and amine groups [113]. The red colloidal gold solution used in this synthesis was prepared via the reduction of chloroauric acid in the presence of sodium citrate [114]. The milky white solution of oil-filled microparticle generated from sonication was poured into a equal amount of the red colloidal gold solution. The mixture was shaken gently for 25 min and allowed to phase separate. Following phase separation, the supernatant took on a reddish color; an indication that the gold particles had been

transferred to the microparticles. These microparticles were centrifuge-washed twice and still retained their reddish color.

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